Optimizing MRI Protocols
(SNR, CNR, Spatial Accuracy)

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Overview

- Parameters Affecting Image Quality
- Manipulating Image Contrast (Standard)
- Classifications of Pulse Sequences
- Relationships Between Resolution & SNR
- MR Image Speed
- Manipulating Image Contrast (Advanced)

Image Quality

- Image Contrast
- Image Resolution
- Signal-to-Noise Ratio
- Imaging Speed

Intrinsic (Tissue) Parameters

- Tissue Water Density (Proton Density)
- Longitudinal Relaxation Time (T1)
  - Magnetization transfer rate coefficients
- Transverse Relaxation Time (T2)
- Chemical Shift (water vs. fat)
- Tissue Motion
  - Macroscopic (flowing blood)
  - Microscopic (apparent diffusion coefficient)
- Tissue Susceptibility (air, fat, bone, blood)

Extrinsic (Selectable) Parameters

- Magnetic Field Strength
- RF Pulse timing
  - Excitation pulse repetition time (TR)
  - Time to echo (TE) & inter-echo spacing (Tₑₑ)
  - Inversion time (TI)
- RF pulse amplitude (flip angles)
- Gradient Amplitude & Timing
  - b-value
- RF pulse excitation frequency & bandwidth
- Receiver bandwidth

Image Contrast

- Basic image contrast is effected by the amplitude and timing of the RF pulses used to excite the spin system.
- More advanced methods may use gradient pulses (to modulate motion) and alter tissue properties with exogenous contrast agents.
Spin (Proton) Density

<table>
<thead>
<tr>
<th>Tissue</th>
<th>% Water</th>
<th>Tissue</th>
<th>% Water</th>
</tr>
</thead>
<tbody>
<tr>
<td>Liver</td>
<td>70%</td>
<td>Spleen</td>
<td>78%</td>
</tr>
<tr>
<td>Kidney</td>
<td>75%</td>
<td>Brain (GM)</td>
<td>85%</td>
</tr>
<tr>
<td>Muscle</td>
<td>73%</td>
<td>Brain (WM)</td>
<td>75%</td>
</tr>
<tr>
<td>Heart</td>
<td>78%</td>
<td>Tendon</td>
<td>60%</td>
</tr>
</tbody>
</table>


Spin Echo Pulse Sequence

<table>
<thead>
<tr>
<th>RX</th>
<th>TX</th>
</tr>
</thead>
<tbody>
<tr>
<td>FID</td>
<td>RF$_1$ (excitation)</td>
</tr>
<tr>
<td>Spin Echo</td>
<td>time</td>
</tr>
<tr>
<td>TE</td>
<td></td>
</tr>
</tbody>
</table>

Out of Phase Spins (Transverse Magnetization)

Spins in Phase  
Precessing  
Out of Phase  

Off-resonance by 17,040 Hz

$17,040 \text{ Hz} = 17,040 \text{ cycles per second} = 360 \times 17,040 \text{ degrees/sec}$

$180^\circ = (360^\circ \times 17,040 \text{ degrees/sec}) \times 3 \times 10^{-5} \text{ s} = 30\mu\text{s}$

Dephasing of MR Signal (Transverse Magnetization)

Coherent Signal  
Less Signal  
No Signal

Signal decays exponentially:

$(\text{after } 90^\circ \text{ rf pulse}) \quad M_{xy} = M_0 \exp \left[-\frac{\text{TE} \cdot T_2^*}{T_2^*}\right]$

where

$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{\gamma}{2\pi} \Delta B_0$ (Magnetic field inhomogeneity)

Rephasing of the MR Signal (Spin Echoes)

Signal Dephasing  
$180^\circ$ pulse  
Rephased Signal

A rephasing $180^\circ$ radio frequency pulse, changes the phase of the net magnetization vectors by $180^\circ$. Thus spins precessing in the same direction come together of moving apart. This forms a SPIN ECHO.

Longitudinal ($T_1$) Relaxation

Longitudinal relaxation is a recovery process:

After $90^\circ$ pulse - all spin energy is kinetic

Spins lose energy to their environment, signal decreases

After $5^\circ T_1$, full potential of net magnetization is reestablished.
Comparison of $T_1$, $T_2$ and for Various Tissues

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$T_1$(ms)</th>
<th>$T_2$(ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Liver</td>
<td>675±142</td>
<td>54±8</td>
</tr>
<tr>
<td>Kidney</td>
<td>559±10</td>
<td>84±8</td>
</tr>
<tr>
<td>Muscle</td>
<td>1123±119</td>
<td>43±4</td>
</tr>
<tr>
<td>Gray Matter</td>
<td>1136±91</td>
<td>87±15</td>
</tr>
<tr>
<td>White Matter</td>
<td>889±30</td>
<td>86±1.5</td>
</tr>
</tbody>
</table>

Akber, 1996 (at 63 MHz)

Time Scale of MR Relaxation Processes

Manipulating Contrast
- The “weighting” of image contrast
- Gradient Echo – manipulating image contrast by varying the flip angle
- Spin Echo – manipulating image contrast with 180° refocusing pulses
- Inversion Recovery – manipulating image contrast with 180° inversion pulses

Rules of Thumb
- TE controls $T_2$ dependence
- TR controls $T_1$ dependence
- $T_1$ competes with $T_2$ and proton density
  "PD Weighted" = long TR, short TE
  "T1 Weighted" = short TR, short TE
  "T2 Weighted" = long TR, long TE

Pulse Sequence Classifications

<table>
<thead>
<tr>
<th>Name</th>
<th>RF Pulses</th>
<th>Contrast Weighting</th>
<th>Application</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gradient Echo</td>
<td>One</td>
<td>T1 or T2</td>
<td>Fast imaging (3DFT)</td>
</tr>
<tr>
<td>Spin Echo</td>
<td>Two or more</td>
<td>T1, PD or T2</td>
<td>conventional</td>
</tr>
<tr>
<td>Inversion Recovery</td>
<td>Three</td>
<td>T1 and T2</td>
<td>Excludes certain tissues</td>
</tr>
</tbody>
</table>

Spin-Echo Imaging Sequence
**Spin Echo Imaging – Effect of TR**

![Spin Echo Imaging](image)

**Multi-Echo Acquisitions**

![Multi-Echo Acquisitions](image)

**Multi-echo Image Matrix**

<table>
<thead>
<tr>
<th>TE (ms)</th>
<th>30</th>
<th>60</th>
<th>90</th>
<th>120</th>
<th>150</th>
<th>180 ms</th>
</tr>
</thead>
<tbody>
<tr>
<td>TR (s)</td>
<td>4</td>
<td>2</td>
<td>1.5</td>
<td>1.2</td>
<td>0.9</td>
<td>0.6</td>
</tr>
</tbody>
</table>

- Most T₁ Weighted
- Most Proton Density Weighted
- Most T₂ Weighted

**Sensitivity Plots** (pulse sequence optimization)

![Sensitivity Plots](image)

**Inversion Recovery**

![Inversion Recovery](image)

- Increased efficiency by scanning multiple planes during TR interval

STIR
(Short TI Inversion Recovery)

Fat
Liver

Inversion Recovery FSE

TR = 1400
TI = 100

TR = 1400
TI = 280

TR = 2000
TI = 280

Modulus
Phase
Compensated

STIR
Fat
Liver

Modulus
Phase
Compensated

FLAIR
(FLuid Attenuated Inversion Recovery)

Brain
Lesion
CSF

TR = 2350
TE = 30

TR = 2350
TE = 80

TR = 2600
TE = 145
TI = 11,000

FLAIR
TR = 2600
TE = 145
TI = 11,000

Multiple Sclerosis

Proton Density
T2-Weighted
FLAIR

Multiple Sclerosis
Proton Density
T2-Weighted
FLAIR

Steady-State Free Precession
(SSFP or True FISP)

SSFP
FISP
FLASH

Relative Sensitivity of Gradient Echo Sequences

SSFP
FISP
FLASH

Also called FAST, ROAST


Vlaardingerbroek & den Il, 1999
Gradient-Echo Imaging

Reference: Wehrli, Fast-Scan Magnetic Resonance. Principles and Applications

Cystic Mass

FISP
TR/TE = 22/10
NSA = 8
6 mm

True FISP
TR/TE = 16/6
flip = 40°
NSA = 8
5 mm

True FISP
TR/TE = 16/6
flip = 70°
NSA = 8
5 mm

Signal-to-Noise Ratio

• Signal-to-noise in an RF coil:

\[ \frac{S}{N} \propto \frac{\omega_0 M_0}{\sqrt{4kT \cdot BW \cdot R_i}} \]

- \( S \) = signal
- \( N \) = noise
- \( \omega_0 \) = resonant frequency
- \( M_0 \) = net magnetization
- \( k \) = Boltzman's constant
- \( T \) = patient temperature
- \( B_1(\vec{r}) \) = RF field distribution
- \( R_i \) = total resistance (coil + patient)

SNR and Imaging Parameters

\[ SNR \propto \frac{FOV_{ro}}{\sqrt{M_{ro}}} \cdot \frac{FOV_{ph}}{\sqrt{M_{ph}}} \cdot \Delta z \cdot \frac{\sqrt{NSA}}{\sqrt{BW_{rx}}} \]

- FOV = field of view
- \( M \) = matrix size
- \( n \) = number of signals averaged
- \( BW_{rx} \) = receiver bandwidth
- \( ro \) = read out (frequency encoding) direction
- \( \Delta z \) = slice thickness
- \( ph \) = phase encoding direction

Matched Bandwidth

• Bandwidth on ADC\(_1\) is limited by finite \( T_{max} \) & desire to minimize \( TE_1 \).

• \( T_1 \) weighting and SNR is sacrificed more by \( T_2 \) decay than by BW.

• \( T_2 \)-weighted image - lots of time for ADC\(_2\), poorer SNR. It would be nice to decrease BW (increase \( T_{max} \)).

Matched Bandwidth = Unmatched Distortion

- Smaller BW on T2W image leads to increased image distortion

Chemical Shift

- Chemical shift (fat-water) ~3.5 ppm
- At 1.5T:
  \[ \Delta f_{\text{fat-water}} = 3.5 \times 10^6 \times \frac{42.6 \text{MHz}}{T} \times 1.5T \approx 220 \text{ Hz} \]
- At 0.5T:
  \[ \Delta f_{\text{fat-water}} = 3.5 \times 10^6 \times \frac{42.6 \text{MHz}}{T} \times 0.5T \approx 73 \text{ Hz} \]
- ↓ field strength .... ↓ chemical shift

Chemical Shift Artifact

- Occurs in
  - Readout direction
  - Phase encode direction
- Controlled by
  - Fat Pre Saturation
  - STIR sequence

Contrast-to-Noise Ratio

\[ CNR = \sqrt{n} \left( \frac{I_a - I_b}{\sigma} \right) \]

\( \sigma \) = standard deviation of the background signal (the noise)
\( n \) = number of signals acquired
\( I_a, I_b \) = signal intensities from tissues “a” and “b”

Steady state free precession (SSFP)

- FIESTA (fast imaging employing steady state acquisition)
- Balanced FFE (fast field echo)
- True FISP (fast imaging with steady precession)

IR Turbo-FLASH

- 180° FID Train (one for each image line)
- T1, TR

Hinton et al. Invest Radiol 2003; 38:436
### 3D FLASH vs. 3D Turbo FLASH

<table>
<thead>
<tr>
<th>IR Turbo FLASH</th>
<th>FLASH</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ti = 500 ms, TR/TE = 10/4.5</td>
<td>TR/TE = 12/4.5</td>
</tr>
<tr>
<td>NSA = 1, flip = 10°</td>
<td>NSA = 1, flip = 8°</td>
</tr>
</tbody>
</table>

### FSE Pulse Sequence

- **RF**
- **G_p**
- **G_m**
- **Overall Signal**
- **Echo**
- **ETL = Echo Train Length**

### Things That have “Blurry” point spread functions are “Sharp” in the Image

K-space  \( \xrightarrow{\text{FT}} \) Image Space

### FSE Point Spread Function Depends on \( T_2 \)

### Effect of Echo Spacing

Signal-to-noise decreases for short \( T_2 \) tissues (gray & white matter) leading to a decrease in spatial resolution.

Vlaardingerbroek & den Boer, 1999

### Multi-Shot EPI Pulse Sequences
Comparison of EPI and SE

T2-weighted SE
TR 2400, TE 80
192x256; 7:44

Single shot EPI
TE = 80
128x128; 120 ms

Nine-shot EPI
TR 3000, TE 21
252x256; 1:06

Magnetization Transfer

PROTON SPECTRUM

Frequency (Hertz)

"Free" Water

Lipids

"Bound" Water

Frequency (Hertz)

Gradient Echo with MTC Pulse

TX
RF excitation

RX
Field Echo

Slic Select

Spoilers or
Crushers

Gsl
Dephasing

Gro
Phase Encode

Gpe

Magnetization Transfer Contrast

Multislice FSE:
Magnetization Transfer Contrast
Enhances T2 Weighted Appearance

Slice thickness - RF bandwidth

- thinner slices require longer RF pulses

\[ \text{Bandwidth} = \frac{1}{\Delta z \Delta \omega} \]

Crafted RF Pulses

\( A \text{sinc function} (\frac{\sin x}{x}) \) envelope on the r.f. pulse produces a nearly square excitation profile of the phantom...
MRI Slice Thickness

- Signal ramps have a slope of 10:1
- Signal from ramp is 10 x slice thickness
- Two ramps are used to compensate for in-plane rotation of the phantom
- Phantom does not compensate for tilting backwards or swaying left-right

Cranial Nerves

Frequency Encoding & FOV

- The field of view (FOV) is determined by the range of frequencies in the x-axis which are sampled:
  \[
  FOV_x = \frac{Bandwidth}{\gamma \cdot G_x}
  \]
- The receiver bandwidth is not the same as the RF bandwidth:
  - ↑ BW ↑ FOV_x
  - ↓ G_x ↑ FOV_x
- FOV must be sufficient to cover all tissue in the slice to avoid aliasing

Phase Encode Gradient

The maximum phase-encoding gradient is:

\[
G_{ph-max} = \frac{2\pi}{\gamma} \cdot BW \cdot FOV = \pi \cdot \frac{N}{\gamma \cdot FOV \cdot T_y}
\]

where FOV is the desired field of view.

The phase-encoding gradient increment is:

\[
\Delta G_{ph} = \frac{\pi}{\gamma} \cdot \frac{1}{FOV \cdot T_y}
\]
- Note that a factor of two is gone from \( G_{ph-max} \) because \( G_{ph} \) runs from \( G_{ph-max} \) to \( G_{ph-max} \)

Sampling the Signal

- Each signal is digitally sampled to produce one line in the k-space matrix:
  \[
  \Delta k_x = \gamma \cdot G_x \cdot \Delta t_x
  \]
  \[
  \Delta k_y = MHz \cdot \frac{mT}{m} \cdot \text{msec}
  \]
  \[
  \Delta k_z = \text{cycles} \cdot \frac{1}{m}
  \]
  \[
  \Delta k_y = \frac{1}{FOV_y}
  \]

CHESS

- This is typically accomplished by preceding a SE or FSE sequence with a 90° pulse that is frequency, not spatially, selective.
Fat Suppression

T1W images without (left) and with (right) fat suppression.

Spatial Saturation

90° Sat Pulse
90°-180° SE Pulses

Blood Flow

Spatial Saturation

a. No Sat Pulse
b. Inferior Sat Pulse
c. Superior Sat Pulse
d. Inferior and Superior Sat Pulses

Gadolinium Contrast Agents

• GD-DTPA, Gd-DOTA, Gd-HP-DO3A:
  the old standards, generally diffusable but not in brain due to Blood Brain Barrier
  – In low doses decrease T1
  – In higher doses decrease T1 and T2*

NEW INTRAVASCULAR AGENTS -
• Blood pool agents
  – Magnetic iron oxide particles
  – Bind serum albumin, have higher relaxivity

Gadolinium Contrast Agents

T1-weighted Images

Normal

With Gd Contrast

Dynamic Contrast Studies

• Dynamic study of breast tumor
• Curves denote difference between uptake of normal glandular tissue & lesion

Vlaardingerbroek & den Boer, 1999

G.D. Clarke, AAPM 2004
Attenuation Due to Diffusion

\[ A(TE) = A(0) \exp[-\gamma^2 G^2 D_{\text{app}} \delta^2 (\Delta - \frac{\delta}{4})] \]

Where:
- \( \alpha = \frac{\pi}{2} \);
- \( G \) is amplitude of diffusion sensitive gradient pulse;
- \( \delta \) is duration of diffusion sensitive gradient;
- \( \Delta \) is time between diffusion sensitive gradient pulses;
- \( D_{\text{app}} \) is the apparent diffusion coefficient

Diffusion Weighted Imaging: Prototype Pulse Sequence

\[ b = \gamma^2 G^2 \delta^2 (\Delta - \frac{\delta}{3}) \]

Stejskal EO & Tanner JE, 1965. 42: 288-292

The b-value

- Controls amount of diffusion weighting in image
- The greater the b-value the greater the area under the diffusion-weighted gradient pulses
  - longer TE
  - stronger and faster ramping the gradients

Diffusion EPI Pulse Sequence

Diffusion Imaging of Infarcts

SE Echo Planar TE = 80 ms
SE Echo Planar b = 1205 s mm\(^{-1}\)

Biorelaxometry

Koenig, et. al, 1983
Specific Absorption Rate (SAR)

- The patient is in an RF magnetic field that causes spin excitation (the B1 field)
- The RF field can induce small currents in the electrically conductive patient which result in energy being absorbed.
- The RF power absorbed by the body is called the specific absorption rate (SAR)
- SAR has units of watts absorbed per kg of patient
- If the SAR exceeds the thermal regulation capacity the patient’s body temperature will rise.

Scan Parameters Effecting SAR

- **Patient size**: SAR increases as the patient size increases – directly related to patient radius
- **Resonant frequency**: SAR increases with the square of the Larmor frequency ($\omega_0$)
- **RF pulse flip angle**: SAR increases as the square of the flip angle ($\alpha$)
- **Number of RF pulses**: SAR increases with the number of RF pulses in a given time

SAR Effects on Pulse Sequences

- Decrease number of slices per study
- Requires decreased flip angles, even in gradient echo sequences
- Forces increases in TR
- Limits use of Fast Spin Echo imaging

Summary

- RF pulses can be crafted to select slices
- RF pulse timing (TE, TR & TI) is used to manipulate soft tissue contrast in MR images
- Gradient echo imaging with partial flip angles is faster than spin echo imaging
- Spin Echoes correct for poor magnetic fields
- Inversion recovery can be used to get rid of fat or CSF from images
- Contrast agents are used to manipulate tissue $T_1$

Uniformity at 3 Tesla

- B1 field maps in a conductive saline phantom (18 cm diameter)

Summary

- Presaturation schemes can effect flow, chemical shift and T1-weighted contrast
- Fast imaging methods are often applied to imaging physiology as well as anatomy
- Typically, SNR is reduced with increases in resolution, contrast and imaging speed
- Various image parameters will have to be adjusted for each magnetic field strength